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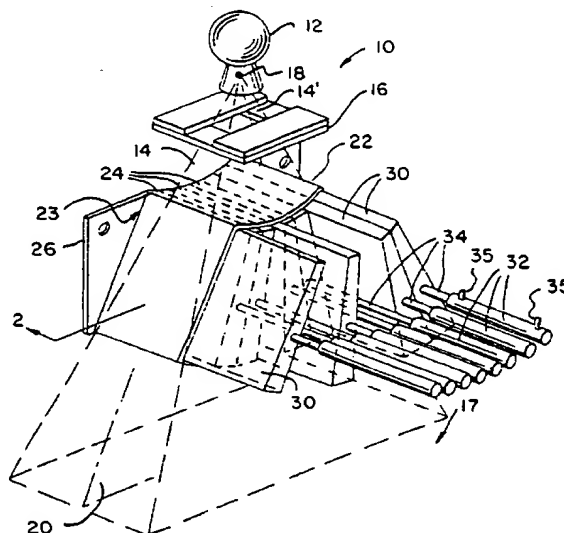
(57) Abstract

A radiation therapy apparatus includes a compensator having a plurality of leaves to divide a radiation beam into rays each of which may be effectively and individually attenuated over a range of attenuations by controlling the ratio of time in which each leaf is in the open and closed states. The compensator is used to compensate for the non-uniform source fluence profile of the radiation beam. In a second embodiment, a radiation therapy apparatus includes an oscillating means for oscillating an electron stream along a line on a radiation generating target so that the resulting field of radiation emanating from the target within the plane of oscillation is uniform over time. The invention eliminates the need for beam flattening filters and thus more effectively uses the energy of the radiation beam.

DYNAMIC BEAM FLATTENING APPARATUS
FOR RADIATION THERAPY

Background Of The Invention

Field of the Invention

This invention relates generally to radiation therapy equipment for the treatment of tumors, or the like, and specifically to a method and apparatus for accommodating
5   uneven fluence within a radiation beam.

Description Of The Art

Medical equipment for radiation therapy treats tumorous tissue with high energy radiation. The radiation dose and the placement of the dose must be accurately
10   controlled to insure both that the tumor receives sufficient radiation to be destroyed, and that damage to the surrounding and adjacent non-tumorous tissue is minimized.

One radiation source employed in radiation therapy is
15   a linear accelerator (linac). A linear accelerator consists of a series of metal electrodes spaced along an evacuated straight path. Every other electrode is connected together to one terminal of a high frequency electric generator, and the intervening electrodes are
20   connected to the opposite terminal of the generator to produce an alternating electric field which accelerates electrons from an electron source. Effectively, an electromagnetic wave is set up that travels continuously through the evacuated chamber and electrons "ride" the
25   electromagnetic wave.

Referring to Fig. 3(a), the electrons accelerated by the linear accelerator are directed at a target 2. When electrons strike one side of the target 2 and pass through

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or near the electric field of a target atom they lose some
energy. Some of the energy lost by an electron may appear
as a radiation photon having a specific quantum of energy.
After each energy loss the electrons are typically

5 deflected in a new direction and may experience several
additional decelerations before finally stopping. Each
deceleration generates radiation photons of widely
different energies. The resulting radiation emitted from
the target thus has a continuous spectrum of energies

10 ranging from zero to the maximum kinetic energy of the
electrons entering the target 2.

The process of slowing an electron and emitting a
photon is called bremsstrahlung or bremssradiation, a
German word meaning breaking radiation because it is

15 associated with a slowing of the electrons. The greater
the initial velocity of the electron and the closer the
electron approaches the nucleus, the more kinetic energy
will be transferred to the bremsstrahlung (i.e. the higher
the energy of the emitted photon).

20 Again referring to Fig. 3(a), the linac directs the
electron beam toward one locus on a target and, a
relatively large radiation fluence emanates from the
opposite side of the target 2 along a central ray 28
concentric with the electron beam. Subsequent electron

25 collisions are normally not along this central ray 28 and
therefore x-ray flux from later collisions is emitted from
the target at angles removed from the central ray 28. The
net effect is that photons emanate from the target at
different angles relative to the central ray 28, a

30 proportionally larger number emanating along the central
ray and a lesser number at angles diverging about this ray
28. A non-uniform radiation fluence profile 100 results.

The radiation beam does not emanate from a single

point on the target but rather from an "emitting area" of
the target. As a result, the non-uniformity of the fluence
profile emanating from the target is accentuated when the
beam is collimated. In such collimation a central portion
5   of the beam is produced in the space traversed by radiation
emanating from all parts of the emitting area. An outer
penumbra is produced in the space traversed by radiation
emanating from only part of the emitting area, (i.e.
radiation passing through the collimator at a relatively
10   sharp angle). The penumbra has less fluence.

The combined effect of the penumbra and the fluence
profile is that the variation of fluence across the beam
field is more extreme.

Radiation therapy planning techniques generally assume
15   that the fluence of the radiation source is uniform
throughout the cross section of the radiation beam.
However, as noted above with respect to linacs, most
sources of radiation produce beams with greater fluence
toward their centers. Accordingly, it is typical to use a
20   beam flattening filter, thicker toward the center, to
render the fluence profile of the beam more uniform.

Referring to Fig. 3(b), a beam-flattening filter
constructed of a radiation attenuating material is placed
directly within a beam to absorb some of the fluence within
25   the central section of a source fluence profile reducing
this central section to a fluence comparable to the fluence
at the edges of the beam. In this manner, the beam fluence
can be controlled and made more uniform.

A drawback to beam flattening filters is that they
30   waste radiation energy, reducing the intensity of the beam
they filter. Therapy sessions with reduced beam
intensities require longer irradiation sessions to deliver
desired irradiation doses to a tumor site and hence limit

therapy throughput rates.

In addition, beam flattening filters also remove low
energy photons from the center of the beam and allow them
to pass around the periphery. This effect is called beam
5   hardening and results in a non-uniform energy spectrum
within the beam.

Further, flattening filters cannot compensate for non-
uniformities that evolve with time and use. Therefore,
therapy planning that relies on assumptions of uniform
10   fluence in the radiation beam may be inaccurate.

Summary of the Invention

The present invention is a method of correcting for
non-uniform fluence profiles in a radiation source used for
radiation therapy that minimizes the amount of wasted
15   radiation energy.

In one embodiment, a dose calculator receives both a
desired fluence profile for the patient and a non-uniform
source fluence profile and generates an attenuation profile
in accordance with the desired fluence profile such that
20   the attenuation profile corrects for the non-uniformity of
the non-uniform source profile. A compensator receives the
desired dose profile and attenuates the non-uniform source
fluence profile to produce the desired fluence profile.

It is thus one object of the invention to provide a
25   simple and reliable means for producing a desired fluence
profile from a radiation source having an arbitrary non-
uniform source profile. The control signal for the
compensator may be adjusted to account for non-uniform
radiation source profiles, using that non-uniformity if it
30   comports with the desired profile and modifying it only as
necessary in other cases. Thus, there is no unnecessary
attenuation as might be produced by a static beam

flattening filter.

The compensator may include a number of radiation
attenuating leaves in a rack positioned within a radiation
beam before it enters a patient. The leaves may move into
5 the radiation beam in a closed state (each leaf thus
occluding one ray of the beam) and move out of the
radiation beam in an open state to allow unobstructed
passage of the associated ray.

A motivation mechanism allows the leaves to be
10 independently moved between the open and closed states and
a timer communicating with the motivation mechanism
controls the ratio of the period of time during which each
leaf is in the closed state to the period in which each
leaf is in the open state to control the average intensity
15 of each ray of the beam.

It is another object of the invention to provide a
method of attenuating the individual rays of a high energy
radiation beam with negligible affect on the spectrum of
the radiation. In contrast, static beam flattening filters
20 attenuate low energy photons yet pass high energy photons
resulting in "beam hardening". The leaves of the
compensator do not attenuate but provide nearly complete
blockage or complete passage without affecting spectral
composition.

25 It is yet another object of the invention to provide a
therapy apparatus that facilitates short irradiation
sessions. By minimally limiting ray intensity and
eliminating a static flattening filter, relatively large
irradiation doses may be directed toward a tumor site in a
30 reduced period of time. Short irradiation sessions mean
minimal patient discomfort and greater throughput rate for
each radiation therapy machine.

In a second embodiment, a flat beam is produced by

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oscillating the electron stream of the linac at a constant
velocity so that the electron stream strikes a radiation
producing target at different points along an impact line
rather than at one point. By oscillating the electron

5   stream, the source fluence profile is "smeared" over the
beam width producing a beam that, although instantaneously
non-uniform, has an average uniformity over time within the
plane of oscillation.

The second embodiment may include a linac with an
10   electromagnet capable of deflecting an electron stream
along a plane of oscillation as it passes through the
electromagnet. By reciprocating the electron stream along
an impact line on the target, the resulting field on the
opposite side of the target has an average uniform

15   intensity along the plane of oscillation. Again, as with
the first embodiment, there is no "beam hardening" (i.e.
each ray of the average uniform profile has the same
spectrum of photons).

Importantly, with this second embodiment, small
20   discrepancies in the source profile along the plane of
oscillation due to either design/defect or worn hardware
are unimportant. The oscillating source profile
effectively compensates for such discrepancies by averaging
the fluence variations over the plane of oscillation. For

25   therapy planning purposes, when the second embodiment of
the compensator is employed to produce an average uniform
profile, the precise source profile of the radiation source
need no longer be known. It is sufficient, for therapy
planning purposes, that the average fluence across the

30   oscillation plane is known.

The foregoing and other objects and advantages of the
invention will appear from the following description. In
the description, reference is made to the accompanying

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drawings which form a part hereof and in which there is
shown by way of illustration several preferred embodiments
of the invention. Such embodiments do not necessarily
represent the full scope of the invention, however, and
5   reference must be made therefore to the claims herein for
interpreting the scope of the invention.

Brief Description Of The Drawings

Fig. 1 is a perspective view of the compensator
assembly used in the present invention positioned within a
10   fan beam of radiation, showing the compensator leaves and
their associated pneumatic cylinders.

Fig. 2 is a schematic view of a radiation source
consisting of a linear accelerator and an accelerator
target;

15   Figs. 3(a) - 3(d) are graphical representations of a
radiation source fluence profile where Fig. 3(a) and 3(b)
show a three and two dimensional uncompensated beam fluence
profile, respectively, and Fig. 3(c) and 3(d) shows the
effects of a beam flattening filter, in three and two
20   dimensions respectively;

Fig. 4 is a cross-section of the compensator assembly
of Fig. 1 along line 4-4 showing the trapezoidal aspect of
each compensator leaf, the fan beam of radiation, and the
guide rails for supporting the compensator leaves when they
25   move;

Fig. 5 is a cutaway perspective view of a set of guide
rails and one leaf of Fig. 4 showing a collar for
supporting the leaf in its fully closed position;

Fig. 6 is a block diagram showing the elements of a
30   radiation therapy apparatus incorporating a computer
controller compensator used in the first embodiment of the
present invention;

Fig. 7 is a block diagram depicting a difference calculator which compares a source fluence profile and a desired fluence profile and generates an attenuation signal;

5   Figs. 8(a)-(d) are dose distributions of a hypothetical tumorous region showing dose intensity by lines of equal dose, with Fig. 8(a) showing a desired dose distribution and Figs. 8(b), (c), and (d) showing progressive actual dose distributions after two, three and

10   ten iterations of a radiation planning procedure used with the present invention;

Fig. 9 is a diagrammatic. representation of a patient receiving radiation therapy, showing the scatter kernel and the coordinate system used to describe radiation planning

15   used with the present invention;

Fig. 10 is a perspective representation of a monodirectional scatter kernel associated with a radiation beam at one gantry angle;

Fig. 11 is a perspective representation of a composite

20   multidirectional scatter kernel associated with a plurality of radiation beams at multiple gantry angles;

Fig. 12 is a block diagram depicting the fluence profile calculator which takes a desired dose map and calculates a fluence profile;

25   Fig. 13 is a block diagram depicting the overall iterative method of controlling the compensator of the present invention, employing the fluence profile calculation method of Fig. 12;

Figs. 14(a)-(c) are perspective views of plots showing

30   the error between the desired dose distribution and the actual dose distribution obtained with the present invention for one, two and four steps of iteration respectively;

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Figs. 15(a)-(b) represent simplified fluence profiles
where Fig. 15(a) is a non-uniform source profile and Fig.
15(b) is a desired profile;

Fig. 16 is a schematic view of a radiation source
5   showing a linac, a deflection yoke, and a target used in a
second embodiment and the compensator of the first
embodiment;

Fig. 17 is a block diagram similar to Fig 6 showing
the elements of a radiation therapy apparatus used in a
10   second embodiment of the present invention;

Fig. 18 is a graphical representation showing the
source fluence profile in a plurality of positions as it is
swept through a plane of oscillation to produce an averaged
uniform profile.

15   Detailed Description Of The Preferred Embodiment
Referring to Fig. 1, a radiation therapy unit 10
suitable for use with the present invention includes a
radiation source 12 producing a generally conical radiation
beam 14' emanating from a focal spot 18 and directed
20   towards a patient 17 (not shown in Fig. 1). The conical
beam 14' is collimated by a radiation opaque mask 16
constructed of a set of rectangular collimator blades to
form a generally planar fan beam 14 centered about a fan
beam plane 20.

25   I.   The Radiation Source
Referring to Fig. 2, the radiation source used in the
present invention is a linear accelerator 1 (linac) used in
combination with an x-ray producing target 2. The linear
accelerator 1 consists of series of metal electrodes 3
30   spaced along a central axis 9 and enclosed in a vacuum.
Alternate electrodes 3a, 3c are connected to one end of a

high frequency alternating current generator 4, and the
intervening electrodes 3b, 3d are connected to the opposite
end of the generator 4. Adjacent electrodes form multiple
radio frequency cavities 5 along the axis 9 and produce
5 fields with the necessary polarization so that electrons 6
between the electrodes 3 will be accelerated along the
central axis 9.

During operation, electrons 6 from a source 7 are
accelerated to a desired speed and directed at the target 2
10 as they pass through the radio frequency cavities 5. When
the electrons 6 impact one side of the target 2, the
impacting electrons 6 lose energy and at the same time
produce x-ray photons 8 that emanate from the opposite side
of the target 2 along the central axis 9.

15   All of the electrons 6 impacting the target 2 do not
continue along the central axis 9. Some electrons 6
scatter within the target 2. After an electron 6 initially
impacts a target atom, the electron 6 loses energy in the
form of a photon, and typically changes direction. The
20 electron 6 continues at a lesser velocity after the first
impact and then typically impacts a second target atom off
the central axis 9 losing additional energy in the form of
a second photon which emanates from the target along the
line of collision rather than along the central axis 9.
25 Electron-atom collisions continue until the electron 6
loses all of its energy, ideally, within the target.

Many photons emanate from the target 2 in directions
diverging about the central axis 9. Thus, as seen in Fig.
3(a), the resulting fluence profile 100 emanating from the
30 target 2 is generally of a bell shape.

Referring to Fig. 16, the linac 1 may employ an achromatic
bending magnet 106 to facilitate a more compact design. As
seen in Fig. 16, by using a bending magnet 106, the actual

position of the linac 1 need not be perpendicular to the x-ray target 2, but may be substantially parallel thereto. Employing an achromatic bending magnet, the actual circumferential width of a gantry 44 used with the
5   radiotherapy system can be limited. Thus, the linac-gantry configuration may be employed in a standard sized radiation therapy room.

The magnet 106 provides desirable achromatic focusing properties. Referring to Fig. 16, the lower energy
10   component 6a of the stream is deflected through a loop of small radius and the high energy component 6b is deflected through a loop of larger radius. The important property of the achromatic magnet is that these components 6a, 6b are brought back together to the same position, angle and beam
15   cross-section at the magnet exit, as they were when they left the linac 1.

II.   The Compensator

Referring again to Fig. 1, a compensator 22 is centered in the fan beam 14 and about a fan beam plane 20,
20   prior to the radiation being received by a patient (not shown in Fig. 1), and includes a plurality of adjacent trapezoidal leaves 30 which together form an arc of constant radius about the focal spot 18. The leaves 30 are held in sleeves 24. The sleeves 24 are constructed of
25   radio translucent materials and attached at their inner ends 23 to a mounting plate 26 which is fixed relative to the focal spot 18. The mounting plate 26 is constructed of a sturdy, possibly radiopaque material and is positioned just outside the fan beam 14 to prevent interference with
30   the fan beam 14.

Preferably, the leaves 30 of the compensator 22 subtend the entire fan beam 14 to divide the fan beam 14

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into a set of adjacent slab-like rays 28 at offset angles
φ. Referring to Figs. 4 and 5, each sleeve 24 is open at
its outer end 27 to receive, by sliding, a comparably sized
trapezoidal leaf 30 constructed of a dense, radiopaque
5 material such as lead, tungsten, cerium, tantalum or a
related alloys.

Each leaf 30 may slide completely within its
corresponding sleeve 24 to block the ray 28 associated with
that sleeve 24. When the leaf 30 blocks its corresponding
10 ray 28, it is referred to as being in a "closed state".
The sleeves 24 are of ample length to permit each leaf 30
to slide out of the path of the fan beam 14, so as to leave
its corresponding ray 28 completely unobstructed, and yet
to still be guided by the sleeve 24. In this non-blocking
15 position, a leaf is referred to as being in the "open
state".

Each leaf 30 may be moved between its open and closed
states by means of a corresponding pneumatic cylinder
connected to the leaf 30 by a flexible link 34. The
20 pneumatic cylinders 32 have internal pistons (not shown)
that may be moved at high velocity between the ends of the
cylinders 32 by means of pressurized air coupled to the
cylinders 32 through supply hoses 35. The supply hoses 35
are fed by a compensator control (not shown in Figs. 1 or
25 4) to be described below. The pneumatic cylinders 32 are
capable of applying high forces to the leaves 30 to move
them rapidly and independently between the open and closed
states.

Referring now to Figs. 4 and 5, the leaves 30 are
30 supported and guided within the sleeves 24 by guide rails
36 fitted into notches 38 cut along the edges of the leaves
30. The notches 38 allow the guide rails 36 to slidably
retain the leaves 30 within the sleeves 24 during motion

between the open and closed states.

In the closed state, the inner end 40 of each leaf 30 is captured by a rigid collar 42 attached to the mounting plate, which aligns the leaf 30, more accurately than may 5 be done by the guide rails 36, with the mounting plate 26 and hence with the fan beam 14. Whereas the guide rails 36, which are ideally radio translucent, are relatively insubstantial, in contrast, the collar 42, positioned outside the fan beam 14 on the mounting plate 26, need not 10 be radio-translucent and hence is more substantial in construction. A collar (not shown) similar to collar 42, supports each leaf 30 when it is fully in the open state. Because the leaves 30 spend most of their time fully in the open or closed states, they are, at most times, firmly 15 located by a supporting collar 4.

A.   Radiation Therapy Hardware Of The First Embodiment

Referring now to Fig. 6, the radiation source 12 is mounted on a gantry 44, the latter rotating within the fan beam plane 20 about a center of rotation 45 in the patient 20 17 so that the fan beam 14 may irradiate a slice of the patient 17 from a variety of gantry angles θ.

The radiation source 12 is controlled by a radiation control module 48 which turns the radiation beam 14 on or off under the control of a computer 51.

25 A compensator control module 52 provides a source of compressed air and valves to gate that air through supply hoses 35 to control, separately, the pneumatic cylinders 32 to move each of the leaves 30 in and out of its corresponding sleeve 24 and ray 28 (see also Fig. 1). The 30 compensator control module 52 also connects with computer 51 to allow program control of the compensator 22 to be described.

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An exit detector 50' positioned on the gantry 44 opposite the radiation source 12 measures the radiation passing through the patient 17 and is connected to a Data Acquisition System 58 as will be described below.

5   Preferably, the exit detector 50' is comprised of a plurality of detector elements each subtending the same angle as one leaf of the compensator 22.

The compensator control module 52, moves the leaves 30 of the compensator 22 rapidly between their open and closed

10 states to either fully attenuate or to provide no attenuation to each ray 28. The ratio between the closed and open states, or the "duty cycle" for each leaf 30, affects the total energy passed by a given leaf 30 at each gantry angle and thus controls the average fluence of each

15 ray 28. The ability to control the average fluence at each gantry angle permits accurate control of the dose provided by the radiation beam 14 through the irradiated volume of the patient 17 by therapy planning methods to be described below.

20 A tomographic imaging system 11 employing an x-ray source 46 and an opposed detector array 50 may be advantageously mounted on the same gantry 44 as the radiation source 12 to produce a tomographic or slice image of the irradiated slice of the patient 17 prior to

25 radiation therapy for planning purposes.

A gantry control module 54 provides the signals necessary to rotate the gantry 44 and change the angle θ of the fan beam 14, hence to change the position of the radiation source 12, exit detector 50', computed tomography

30 x-ray source 46 and detector array 50. Gantry control module 54 connects with computer 51 so that the gantry may be rotated under computer control and also to provide the computer 51 with a signal indicating the gantry angle θ to

assist in that control.

Control modules for the tomographic imaging system 11
include: x-ray control module 56 for turning on and off the
x-ray source 46, and data acquisition system 58 for
5 receiving data from the detector array 50 in order to
construct a tomographic image.

An image reconstructor 60 typically comprising a high
speed array processor or the like receives the data from
the data acquisition system 58 in order to assist in
10 "reconstructing" a tomographic image from such data
according to methods well known in the art. The image
reconstructor 60 also communicates with computer 51 to
assist in high speed computations used in the present
invention as will be described below. The tomographic
15 image allows verification of the patient setup just prior
to radiation therapy treatment.

A terminal 62 comprising a keyboard and display unit
63 allows an operator to input to programs and data to the
computer 51 and to control the radiation therapy and
20 tomographic imaging equipment 10 and 11 and to display
tomographic images produced by the image reconstructor 60
on the display 63. A mass storage system 64, being either
a magnetic disk unit or tape drive, allows the storage of
data collected by the tomographic imaging system 11 for
25 later use.

Computer programs for operating the radiation therapy
system 10 will generally be stored in mass storage unit 64
and loaded into the internal memory of the computer 51 for
rapid processing during use of the system 10.

30   B.   Operation of the Therapy Equipment
Prior to a radiation therapy session, the radiation
source 12 and exit detector 50' can be used to identify the

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source fluence profile 100 (source profile) of the
particular radiation source 12 (see Fig. 3(a)). The
radiation source 12 may direct a beam toward the exit
detector 50' with no intervening patient 17. The source

5 profile 100 can be obtained and divided into separate beam
rays 28 so that a discrete fluence value can be associated
with each leaf 30 of the compensator 22.

Referring to Fig. 7, once the source profile 100 is
obtained, a dose calculator 102 effectively compares the

10 source profile 100 to a desired fluence profiles 101
(desired profile) to be directed at a patient for every
gantry angle θ. The dose calculator 102 generates duty
cycles 103 indicating the ratio of time each leaf 30 should
be in the closed state to the time each leaf 30 should be

15 in the open state to attenuate the source profile 100 and
produce the desired profile 101.

Using the duty cycles 103, a controller 104 takes into
account the non-uniform source profile 100 and generates an
attenuation profile 105 corresponding to each desired

20 profile 101. The attenuation profiles 105 control the
movement of compensator leaves 30 at every gantry angle θ.
The attenuation profiles 105 directed at an arbitrary slice
of the tumor make up a treatment sinogram for a particular
tumor slice.

25 The attenuation profiles 105 needed to generate the
treatment sinograms are determined by therapy planning
software (described below) and stored in the computer 51.

C. Therapy Planning Software
The generation of a treatment sinogram needed to

30 obtain the full benefits of the above described compensator
is performed by specially developed software running on the
computer 51 and reconstructor 60. Although the treatment

planning is performed in software, it will be recognized
that the planning may also be implemented in discrete
electronic circuitry dedicated to this operation and that
such dedicated circuitry may be employed to provide even
5 greater speed to this process.

Referring to Fig. 8(a), the generation of the desired
treatment sinogram to control compensator 22 begins with
the definition of a desired dose map 66. A typical desired
dose map 66 assigns a relatively high radiation dose,
10 within a dose constraint, to an area of tumorous tissue 68
and a second lower radiation dose to the area of healthy
tissue 70 outside of that region. The healthy tissue 70
may include an area 72 including a radiation sensitive
organ or the like to which an even lower radiation dose may
15 be assigned.

The desired dose map 66 is stored within the memory of
computer 51 as an array of elements each element holding
one digital value, and may be most easily entered by
displaying the tomographic view of the slice of patient 17
20 on the display 63 of the terminal 62 and manually tracing
around the tumorous area 68 using of a track-ball or
similar input device as is well understood in the art.
Standard area-filling computer programs may be used to
transfer the dose values assigned to each traced region to
25 the appropriate element in the array of memory representing
the desired dose map 66.

Each element of the dose map 66 thus defines the dose
desired at each of the plurality of volume elements 74
("voxels") within a slice of the patient 17. Referring to
30 Fig. 9, each voxel 74 of the patient 17 may be identified
by a vector $\vec{r}$ defined from a given reference point 76.
The dose at each voxel 74 is $D(\vec{r})$.

1. Converting Dose to Terma

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a. Terma

Generally, the dose at any voxel $\vec{r}$ will depend on the energy received at that voxel $\vec{r}$ from radiation scattered from adjacent voxels $\vec{r}'$ (where adjacent voxels $\vec{r}'$ include
5   the voxel $\vec{r}$, i.e., the radiation received directly from the radiation source 12). The dose $D(\vec{r})$ for a given voxel $\vec{r}$ is given by the following formula:

$$D(\vec{r}) = \int T(\vec{r}') A(\vec{r}-\vec{r}') d^3\vec{r}' \qquad (1)$$

where $T(\vec{r}')$ is a value indicating the magnitude of the
10   primary total energy released from $\vec{r}'$ per unit mass of that voxel $\vec{r}'$ and is called the "terma" (total energy released per unit mass).

For a monoenergetic external radiation source, the terma rate $\dot{T}(\vec{r})$ is described by:

15
$$\dot{T}(\vec{r}) = \frac{\mu}{\rho}(\vec{r}) E \int \dot{\phi} (\vec{r}') dt \qquad (2)$$

where $\frac{\mu}{\rho}$ is an effective mass attenuation value at the voxel $\vec{r}'$, E is the energy of the radiation photons in Joules, $\dot{\phi}$ is the distribution of the fluence rate (flux density). The integration of energy times fluence rate
20   over time is energy fluence $\Psi(\vec{r}')$ where:

$$\Psi(\vec{r}') = E \int \dot{\phi} (\vec{r}') dt \qquad (3)$$

hence

$$T(\vec{r}') = \frac{\mu}{\rho} (\vec{r}') \Psi (\vec{r}') \qquad (4)$$

Equation (4) basically relates how much energy from the ray 28 interacts with the voxel r'.

b. Convolution Kernel

$A(\vec{r}-\vec{r}')$ is a convolution kernel describing non-
5 stochastic energy transport or scattering in a uniform
medium. $A(\vec{r}-\vec{r}')$ thus describes how the energy from each
voxel $\vec{r}'$ spreads to contribute to the dose at voxel $\vec{r}$.

The kernel $A(\vec{r}-\vec{r}')$ may be generated using a Monte
Carlo method as is generally understood in the art. As
10 mentioned, it is a three-dimensional function indicating
the fraction of energy absorbed at voxel $\vec{r}$ per unit of
energy released at voxel $\vec{r}'$. The energy emitted from the
terma of each voxel $\vec{r}'$ finds it source in a directed ray 28
from external radiation source 12 and thus $A(\vec{r}-\vec{r}')$ is
15 generally anisotropic as suggested in Fig. 10, spreading
outward away from the entry of ray 28. Energy conservation
requires that:

$$\int A(\vec{r}')\,d^3\vec{r}' = 1.0 \tag{5}$$

That is, if the energy transferred by the primary
20 interaction were all deposited on the interaction point,
the kernel would be approximated as a delta function.

Referring still to Fig. 10, the anisotropy of $A(\vec{r}-\vec{r}')$
is related to the gantry angle θ and thus of the angle of
incidence of the ray 28 at $\vec{r}'$. If the gantry angles θ at
25 which the patient 17 is irradiated are predetermined, a
multidirection convolution kernel $B(\vec{r}-\vec{r}')$ shown in Fig. 8,
may be created from weighted superimposition of the kernels
$A(\vec{r}-\vec{r}')$.

Referring to Fig. 11, assuming that the spreading of
30 radiation is approximately equal for all beam directions

20

and the rays 28 from each gantry angle θ contribute equally
to the terma at voxel $\vec{r}'$, then the multidirectional
convolution kernel reduces to a "isotropic" form as
follows:

5
$$B(\vec{r}-\vec{r}') = \frac{1}{n} \sum_{i=1}^{n} A(\vec{r}-\vec{r}')_i \qquad (6)$$

where n is the number of discrete gantry angles from
which rays 28 are projected.

For multiple rays 28 at different gantry angles, the
total dose at a given voxel $\vec{r}$ is the sum of doses from
10   each constituent beam is therefore:

$$D(\vec{r}) = \int T(\vec{r}')B(\vec{r}-\vec{r}')d^3\vec{r}' \qquad (7)$$

where $T(\vec{r}') = nT(\vec{r}')_i$ the latter term being the
contributed portion of the terma for the ith gantry angle.

This simplification assumes that the contribution to
15   the terma from each ray 28 is equivalent and takes
advantage of the distributive property of convolution.
Errors in this assumption are reduced by the filtration to
be discussed later.

Equation (7) substantially simplifies the calculation
20   of dose from terma but still requires a convolution for
each voxel $\vec{r}$ times the total number of voxels $\vec{r}'$ to
calculate the dose over the entire patient volume.
Preferably, therefore, the calculational efficiency of the
fast Fourier transform can be used and equation (7)
25   converted to the following:

$$D(\vec{r}) = F^{-1}\{F\{T(\vec{r}')\} \cdot F\{B(\vec{r}-\vec{r}')\}\} \qquad (8)$$

where F and F^{-1} symbolize Fourier and inverse Fourier
transforms respectively. This simplification of equation
(8) requires that the kernel $B(\vec{r}-\vec{r}')$ be spatially invariant

and relies on the convolution theorem which states that
convolution of two spatially invariant quantities in a
space domain is equivalent to multiplication in the
frequency domain.

5   The assumption of the spatial invariance of $B(\vec{r}-\vec{r}')$ is
correct only to a first order approximation. Typically,
the kernel $B(\vec{r}-\vec{r}')$ for an external radiation source 12 is a
complex function of: (1) beam hardening of a polyenergetic
x-ray beam (i.e., the effect of the filtration of the
10   patient 17 in increasing the proportion of high frequency
or high energy radiation components), (2) the number of
rays 28 crossing each voxel, and (3) exponential
attenuation by the patient mass.

      In the preferred embodiment, this first factor, beam
15   hardening, is neglected because it is an effect smaller
than the attenuation problem. Thus, the photon energy
spectrum in the patient 17 may be assumed to be the same as
that of the external radiation source 12. This
simplification is not required, however, and it will be
20   understood that beam hardening could be accurately
accounted for by representing a photon energy spectrum by a
finite number of separately convolved energy intervals.

      The second factor, the difference in the number and
orientation of rays 28 that cross each voxel, caused by the
25   geometry of a finite number of gantry angles and the fan
orientation of the beam 14, affect spatial invariance.
Problems arising from the fan orientation of the beam (in
contrast to a parallel beam geometry) are largely solved if
there is a full rotation of the gantry 44. Errors
30   resulting from the fact that irradiation is performed at
only a finite number of gantry angles have been determined
to be acceptable.

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The third factor affecting the assumption of spatial
invariance is the attenuation of the medium. This affects
the fractional contribution of the total terma from the
beams at each gantry angle. Accordingly, in those steps of
5   the planning procedure, as will be noted below, where
accurate calculation of dose is critical, the dose
distribution is calculated separately for each beam based
on the attenuation of overlying voxels, such attenuation
deduced from the parameters of the tomographic image. In
10 this case the simplification of equation (8) may not be
employed and repeated convolutions must be performed. In
certain steps in the planning process, however, as will be
noted, an estimate is sufficient and in these cases $B(\vec{r}-\vec{r}')$
is assumed to be spatially invariant and the dose
15 calculated according to equation (8).

Production of terma values from a desired dose map 66
is then simply the process of inverting equation (8) as
follows:

$$T(\vec{r}') = F^{-1}\left\{\frac{F\{D(\vec{r})\}}{F\{B(\vec{r}-\vec{r}')\}}\right\} \qquad (9)$$

20 This inversion requires that there be no significant
"zeros" (typically at high frequencies) in the denominator
term $F\{B(\vec{r}-\vec{r}')\}$ or more simply that the kernel $B(\vec{r}-\vec{r}')$ be
spatially compact (i.e., the Fourier transform of a
spatially compact kernel will have significant high
25 frequency content). It has been determined by the present
inventors that the kernels dictated for patients 17 are
sufficiently compact to allow this Fourier deconvolution.

Referring now to Fig. 12, this deconvolution to
produce a terma map 82, giving the terma for each voxel $\vec{r}$,
30 from the desired dose map 66, is represented by process
block 80.

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2.   Converting Terma to Voxel Energy Fluence

Knowing the terma map 82, the energy fluence $\Psi(\vec{r}')$, which is a measure of the beam intensity, can be determined at each corresponding voxel by equation (4) from a

5   knowledge of μ/ρ as follows:

$$\Psi(\vec{r}') = \frac{T(\vec{r}')}{\frac{\mu}{\rho}(\vec{r}')} \qquad (10)$$

The value of μ/ρ may be estimated and considered a constant or actual μ/ρ may be deduced from the tomographic scan data collected by means of the tomographic imaging

10   system 11, (shown in Fig. 6). In this manner and as illustrated by process block 84 of Fig. 12, a fluence map 86 giving the fluence at each point of the terma map may be determined.

3.   Converting Voxel Energy Fluence to Energy Fluence

15   Profile

The energy fluence $\Psi(\vec{r}')$ at each voxel $\vec{r}'$ is related to the energy of the ray 28 exiting the compensator 22 by the relation:

$$\Psi(\vec{r}') = \Psi_0(\phi,\theta) e^{-\int \frac{\mu}{\rho}(\vec{r})\rho(\vec{r})\delta(p-\hat{r}\cdot\vec{r})d\vec{r}} \left(\frac{SSD^2(\phi,\theta)}{|\vec{r}|^2} \right) \qquad (11)$$

20   where $\Psi_0(\phi,\theta)$ is the energy fluence for a given ray 28 as described by $\delta(p-\hat{r}\cdot\vec{r})$ at the exit of the compensator 22 and serves to define the fluence profile of the compensator and θ and ϕ are the gantry angle and the offset angles of the ray 28 as previously described.

25   The exponential term represents the attenuation of the ray 28 from the exit of the compensator 22 to the voxel $\vec{r}$ caused by the mass of the patient 17 where by $\mu/\rho(\vec{r})$ is the attenuation for each voxel $\vec{r}$ along the ray 28, $\rho(\vec{r})$ is

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the density of each voxel $\vec{r}$, $SSD(\phi,\theta)$ is the distance between the exit of the compensator 22 and the surface of the patient 17 , $\hat{r}$ is a unit vector along $\vec{r}$ (where the origin of $\vec{r}$ is now assumed to be the center of rotation of the gantry 45), and p is the perpendicular distance from the gantry's center of rotation 45 and the ray 28. The vector is simply a vector along the ray 28 to provide an integration variable.

The fluence at each voxel $\vec{r}$ is related to the fluence of the radiation beam 14 emitted from the compensator 22 by equation (11). In the preferred embodiment, the density and attenuation of each voxel $\vec{r}$, $\mu/\rho(\vec{r})$ and $\rho(\vec{r})$ are assumed to be constant and the fan beam of radiation is approximated by a parallel beam, hence $\dfrac{SSD^2(\phi,\theta)}{|\vec{t}|^2} = 1$

Borrowing from the mathematics of tomographic image reconstruction, the fluence map 86 may be "reverse" back projected (i.e. projected) by projector 85 to determine a fluence profile to be produced by the external-source necessary to generate the desired fluence map and hence dose.

This projection is simply the opposite of a typical back projection used to form an image of a tomographic slice of a patient 17 from a series of projections taken in a tomographic imaging system. Because a projection is a line integral across a distribution, the energy fluence distribution for each voxel (equation (11)) is first differentiated with respect to the rayline $\vec{t}$:

$$\frac{d\Psi(\vec{r})}{dt} = \left[\frac{\mu}{\rho}(\vec{r})\,\rho(\vec{r})\,\delta(\rho-\hat{r}\cdot\vec{r})+\frac{2}{t}\right]\Psi(\vec{r}) \qquad (12)$$

The line integral of $\dfrac{d\Psi(\vec{r})}{dt}$ along $\vec{t}$, corrected for

●   ●

attenuation and inverse-square fall off, then represents
the projection operation and $\Psi_0(\phi,\theta)$ the fluence profile
over the offset angles ϕ of each gantry angle θ, is:

$$\Psi_0(\phi,\theta) = \int \left[\frac{\mu}{\phi}(\vec{r}) \, \rho(\vec{r}) \, \delta(p - \hat{p} \cdot \vec{r}) + \frac{2}{t} \right]$$

5 $$\times \left(\psi(\vec{r}) \, e^{+\int \mu/\rho(\vec{r}) \, \rho(\vec{r}) \, \delta(p-\hat{p}\cdot\vec{r}) \, dt} \left(\frac{|\vec{t}|^2}{SSD^2_{(\phi,\theta)}} \right) \right)$$

$$\times \delta(p - \hat{p} \cdot \vec{r}) \, d\vec{t} \tag{13}$$

The projection of equation (13) is represented by
projector 85 in Fig. 12.

The projection process, for the purpose of computing
10 fluence profiles for the compensator 22, differs in a
fundamental way from the simple inverse of tomographic back
projection. The difference is primarily in a concern for
the sharpness in the transition of the dose between the
irradiated tumorous tissue 68 and the healthy tissue 70.
15 Sharpness in this transition region reduces the irradiation
of healthy tissue 70 and is preferred over actual fidelity
of the dose to the desired dose map 66.

For this reason, the fluence map 86 from the fluence
calculator 84 is prefiltered as shown by process block 88
20 to produce a filtered fluence map $\Psi'(\phi,\theta)$ as follows:

$$\Psi'(\phi,\theta) = F^{-1}\{F\{\Psi(\phi,\theta) \, |\omega|\}_+ \tag{14}$$

where $\Psi(\phi,\theta)$ is the fluence map 86 and $|\omega|$ is a ramp
filter in frequency space and the '+' subscript indicates
the positive component of the filtering result. This
25 prefilter 88 serves to increase the high frequency content
of the fluence map 86 and thus to aid in rapid transition
of dose at the tumor/non-tumor interface.

It is noted that this prefilter 88 is similar to the
filter used in tomographic imaging's "filtered" back

projection. That is, like tomographic imaging, the filter
de-emphasizes the low frequency components of the
projection in producing image data. In addition other
prefilters may be applied to correct for the use of the
5   radially symmetric kernel (equation (6)) in computing the
dose map from the terma map composed from the fluence
profile.

$\Psi'(\phi,\theta)$ will be termed the "desired fluence profile
and refer to the desired profile of the radiation beam
10   received by the patient.

In practice the computation of a terma map, the
generation of a fluence map and the calculation of the
fluence profiles need not be performed as discrete steps
but may be accomplished by a direct projection of the dose
15   map with appropriate filtering. The filtering is
accomplished by a "fast inversion filter" which combines in
projection space the operation of deconvolution and ramp
filtration.

This may be symbolically specified by the following
20   equation

$$\Psi(\phi,\theta) = \wp\{D(\vec{r})\} \otimes I(t) \qquad (15)$$

where $\wp$ refers to a projection operation and I(t) is
the fast inversion filter. The $\otimes$ operators refers to a
convolution operation such as would normally be done in
25   Fourier space using a fast Fourier transformation.

Referring still to Fig. 12 the fluence profile
calculations of block 78, including the deconvolver 80, the
fluence calculator 84, the prefilter 88 which includes any
projection space filter (such as a ramp filter, a fast
30   inversion filter followed by truncation of zeros), and the
projector 85 thus produce fluence profiles which together
create an estimated treatment sinogram 87' from the desired
dose map 66. The fluence profile calculator 78 may use the

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Fourier convolution of equation (9) in the estimate of the
fluence profiles at this stage, accepting minor
inaccuracies in that process, to be corrected at a later
stage, as will be described below.

5   4. Iteration
           Referring now to Fig. 13, the fluence profile
calculator 78 converts the desired dose map 66 to an
estimated treatment sinogram 87', however the fluence
profiles of this estimated treatment sinogram 87' may not
10 be used to control the compensator 22 because, in general,
the estimated treatment sinogram 87 will include positive
and negative values of fluence. Only positive values of
fluence are physically realizable by the compensator 22; a
negative value of fluence would represent a ray 28 that
15 absorbed radiation along its path which is physically
unrealizable.

           Accordingly, at process block 88, the fluence values
of the estimated treatment sinogram 87' are truncated to
positive fluence values 89. As a result of this
20 truncation, the estimated treatment sinogram 87' no longer
produces the desired dose map.

           The amount of error resulting from the truncation
producing the positive fluence profiles 89 is determined by
back projecting the positive fluence values 89 to an actual
25 dose map 90 deviating from the desired dose map 66. This
back projection is accomplished by computing a fluence map
from the positive fluence values 89 per equation (11) and a
terma map per equation (4) and then convolving the terma
map with the kernel per equation (7) to establish the
30 actual dose map 90 per process block 92 of Fig. 13.

           In this back projection, the assumption of spatial
invariance of the convolution kernel $B(\vec{r}-\vec{r}')$ is not made so
as to produce a more accurate actual dose map 90.

The projection of a fluence profile onto a patient 17 to compute a dose map may be performed by a number of other procedures known to those of ordinary skill in the art.

The actual dose map 90 is compared to the desired dose
5 map 66 to produce residual dose map 96 as indicated by process block 94. In the preferred embodiment, the comparison subtracts from the values of each voxel $\bar{r}$ of the actual dose map 90, the greater of: a) the corresponding value of desired dose map 66, or b) a
10 predetermined upper dose constraint. The predetermined upper dose constraint is a threshold number that is deemed an acceptable dose to tumorous tissue 68. Clearly, other methods of quantifying the difference between the desired dose map and the actual dose map will be apparent from this
15 description to those of ordinary skill in the art.

The result of this comparison process 94 is to produce a residual dose map 96 shown in Fig 14(a). This residual dose map 96 is then, again, operated on by the fluence profile calculator 78 (in lieu of the desired dose map 66)
20 to produce an error fluence profile 98 (in lieu of the estimated treatment sinogram 87).

A thus produced error fluence profile 98 is subtracted by subtracter 100 from the estimated treatment sinogram 87' to produce a new estimated treatment sinogram 87.

25 As shown in Fig. 13, this new estimated treatment sinogram 87 is repeatedly operated on by process block 88, 92, 94 and 78 for a predetermined number of iterations, the magnitude of the error fluence profile 98 values decreasing with each iteration as shown in Figs. 13(b)-(d) until a
30 suitably low error fluence profile 98 is obtained.

The new estimated fluence profile 87 is then truncated per process block 88 to produce a final fluence profile sinogram 91.

Referring again to Figs. 8(b), (c) and (d), dose maps obtained by the present invention corresponding to a desired dose map 66 of Fig. 8(a) are shown after: one iteration (Fig. 8(b)); two iterations (Fig. 8(c)); and ten
5   iterations (Fig. 8(d)). The variations in dose in the target volume shown in Fig. 8(d) are plus or minus 2% about the predetermined upper limit of 1,000 cGy.

        5.   Creating An Attenuation Profile From The Source Profile and Desired Profile
10   As described above, the source profile 100 can be obtained by running a test in which substantially no mass is placed within the space between the radiation source 12 and the detector array 50'. Referring to Fig. 15(a), a simplified source profile 100 includes six separate rays 28
15   each having a different value of fluence $\Psi(\theta, \phi)_S$ generally being higher for the center rays per the profile 100 of Fig. 3(a). These rays 28 must be attenuated to attain a desired profile 101 of Fig. 15(b). The desired profile 101 may require rays 111 having no intensity and attenuated
20   rays 112 in other areas.

        Knowing the desired profile 101 and the source profile 100, a dose calculator 104 can generate a fraction of open time within the duty cycle for each leaf 30 necessary to modify the source profile 100 to produce the desired
25   profile 101 as follows:

$$\text{Open Fraction of Duty Cycle } (\phi,\theta) = 1 - \left(\frac{\Psi(\phi,\theta)_S - \Psi(\phi,\theta)_D}{\Psi(\phi,\theta)_S} \right)$$

(16)

        Where $\Psi(\phi, \theta)_S$ refers to the fluence of one ray of the source profile 100, $\Psi(\phi, \theta)_D$ refers to the fluence of the
30   corresponding ray of the desired profile 101 where

$\Psi(\phi, \theta)_D \leq \Psi(\phi, \theta)_S$. θ and ϕ are the gantry angle and the offset angles of the ray 28 (each leaf 30 being associated with one ray 28).

Referring to Fig. 7, a controller 104 combines the
5   duty cycles for all of the leaves 30 at a given gantry angle θ to produce an attenuation profile 105 for the gantry angle θ. The attenuation profile 105 can then be stored in the mass storage unit 64 and called upon later to direct the movement of the compensator leaves 30. When the
10   attenuation profiles 105 are used, the non-uniform source profile is accounted for during irradiation and the full radiation capability of the radiation source 12 is used.

IV.   The Second Embodiment Of The Radiation Source

Referring again to Figs. 6 and 17, the hardware
15   configurations of the first and second embodiments of the present invention are substantially the same. The only difference between the two configurations is that the collimator control 52 in Fig. 6 is replaced by a deflection yoke control 116 in Fig. 17 and the collimator 22 in Fig. 6
20   is replaced by a deflection yoke inside the radiation source 12 in Fig. 17.

Referring to Fig. 16, a radiation therapy compensator constructed in accordance with a second embodiment of the present invention has a linac 1 producing an electron
25   stream 6 that is redirected by an achromatic bending magnet 106 toward a target 2. Among the parts unique to the second embodiment is a deflection yoke 108 disposed along the electron path between the linac 1 and the target 2.

The yoke 108 consists of a non-achromatic bending
30   magnet 111 capable of varying the field within its magnetic passageway 112. Electrons 6 accelerated by the linac 1 pass through the magnetic passageway 112 prior to impacting

the target 2. By changing the field strength within the
magnetic passageway 112, the electron stream 6 can be
deflected within a single plane so that the stream 6
impacts the target 2 at different positions along an impact
5   line 120 rather than at a single point. The non-achromatic
yoke 108 may oscillate the electron stream 6 along a plane
of oscillation 109. During such oscillation, the source
profile 113 moves with the point of impact of the electrons
and target 2.

10   Referring to Fig. 18, by oscillating a non-uniform
source profile 113 across a wide field, the resulting
intensity across the field can effectively be made uniform
as the multiple non-uniform source profiles 113 are
averaged, thereby, forming a uniform average profile 114.
15   It should be recognized that irregularities in the source
profile 113 need not be known because they are averaged
into the uniform average profile 114. It should also be
recognized that relatively wide uniform radiation fields
can be generated.

20   In operation, prior to a therapy session, a test can
be run in which no mass is placed between the radiation
source 12 and the detectors 50' thereby generating a
uniform average profile 114 for therapy planning purposes.
           The above description has been that of two embodiments
25   of the present invention. It will occur to those who
practice the art that many modifications may be made
without departing from the spirit and scope of the
invention.
           For example, the fluence profile calculator 78 that is
30   part of the therapy planning software could be used instead
of the dose calculator 102 to take into account the non-
uniform source profile to generate an attenuation profile
for each gantry angle θ weighted to compensate the non-

uniform field of the source fluence. In this manner the
duty cycle calculation (Equation. 16) could be worked into
the fluence calculation at an earlier stage.

It should also be understood that both embodiments may
5   be used together to achieve a controllable "wide" fan beam
field. As seen in Fig. 16, the non-achromatic yoke 108 can
be placed between the achromatic bending magnet 106 and the
compensator 22. The yoke 108 can produce a wide, uniform
field with little loss of energy while the compensator 22
10 may control the intensity of individual rays within the
wide field. In the alternative, the compensators can be
used together, one controlling a first dimension of the
field and the other perpendicular thereto controlling the
second dimension. For example, the non-achromatic yoke 108
15 could flatten the field along field thickness while the
compensator 22 compensates the field along the field width.
Further, it is not imperative that the first embodiment be
used with a linear accelerator. The compensator 22 may
operate in conjunction with any radiation producing source.
20   Further, by varying the sweep speed of the electron
beam of the linac, the fluence of the rays 28 may be
controlled to augment or eliminate the need for the
compensator 22. In order to apprise the public of the
various embodiments that may fall within the scope of the
25 invention, the following claims are made:

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CLAIMS

We claim:

1.   In a radiation therapy machine having a radiation
source for producing a radiation beam directed toward a
patient at a plurality of treatment angles, the beam
including a plurality of adjacent rays spaced about one
5   central ray, each ray having an energy fluence, the energy
fluences together forming a non-uniform source fluence
profile, an attenuation apparatus comprising:

a dose calculator for receiving a desired fluence
profile for the patient and the non-uniform source fluence
10   profile and for producing duty cycles indicative of how the
non-uniform source profile must be altered to produce the
desired fluence profile;

a controller for receiving the duty cycles from the
dose calculator and producing an attenuation profile
15   corresponding to each desired fluence profile; and

a compensator receiving the attenuation profiles for
attenuating rays of the radiation beam to correct the non-
uniform source fluence profile of the radiation beam to
produce the desired fluence profile.

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2.   The apparatus as recited in claim 1 wherein the
compensator comprises:

5   a plurality of radiation attenuating leaves;
           a supporting structure for guiding the leaves between:
                        a closed state centered on
                the rack within the radiation beam,
                each leaf thus occluding one ray of the
10   beam;
                        an open state with each leaf
                displaced outside of the radiation beam
                allowing unobstructed passage of the
                ray;
15   motivation means for independently moving each leaf
       between the open state and the closed state; and
           timing means communicating with the motivation means
       for controlling the ratio of the period of time during
       which each leaf is in the closed state to the period during
20   which each leaf is in the open state to control the average
       energy fluence of each ray of the beam.

3.   In a radiation therapy machine having a radiation
source for producing a fan beam of radiation diverging
along a plane and directed toward a patient, the radiation
5   source comprising:
       a target;
       an electron generator to produce a stream of electrons
directed at the target, the electrons impacting the target
and producing a plurality of beam rays diverging about one
10   central ray, the fluence of the diverging rays describing a
non-uniform fluence profile; and
       an electron deflecting means positioned between the
electron source and the target to oscillate the electron
stream so that the electron stream impacts the target at
15   different positions to produce a radiation beam having a
substantially uniform average fluence profile.

4.   The machine as recited in claim 3 wherein the
electron deflecting means comprises a bending magnet which
magnetically oscillates the electron stream.

5.   The therapy machine as recited in claim 3 wherein
the target is tungsten.

6.   The therapy machine as recited in claim 3 wherein
the magnetic oscillation is along the plane of the fan
beam.

FIG. 1

FIG. 2

HIGH
FREQUENCY
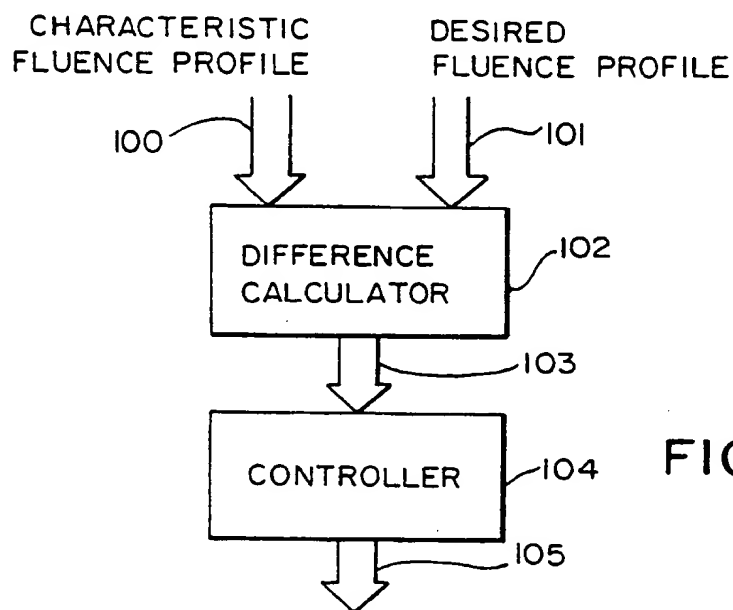AC
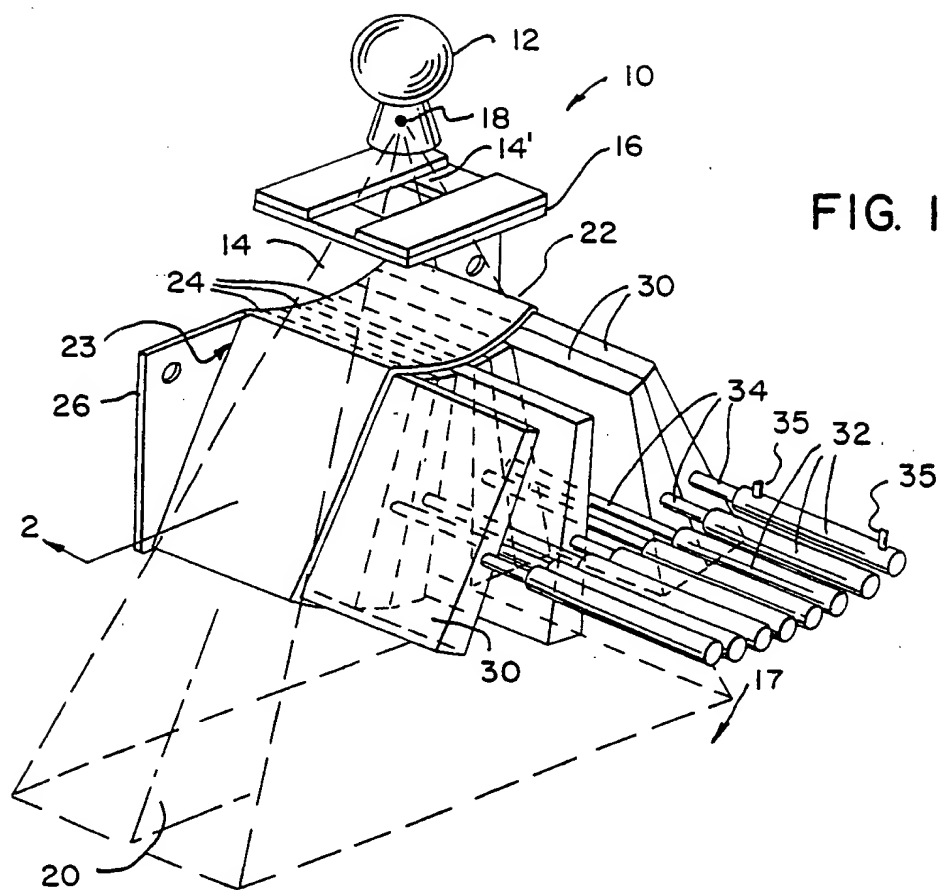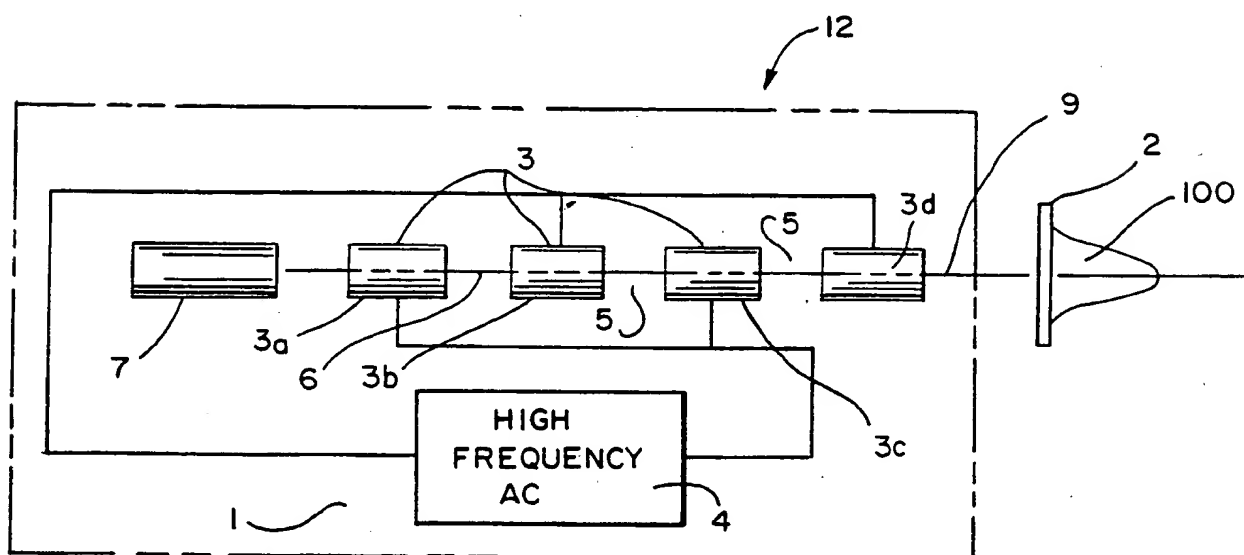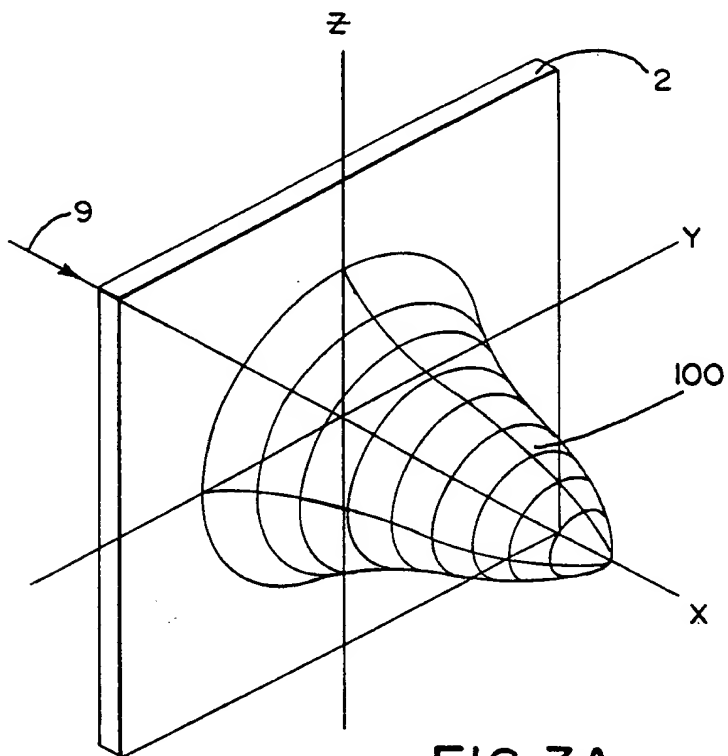
FIG.3A
PRIOR ART
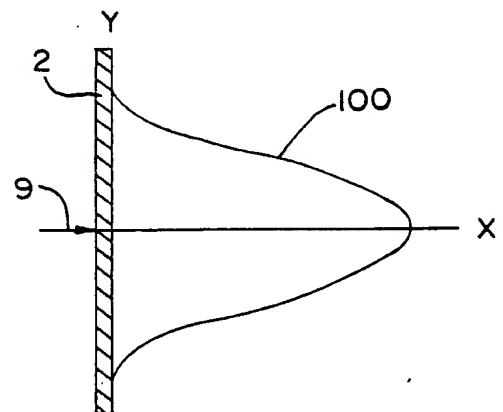
FIG.3B
PRIOR ART
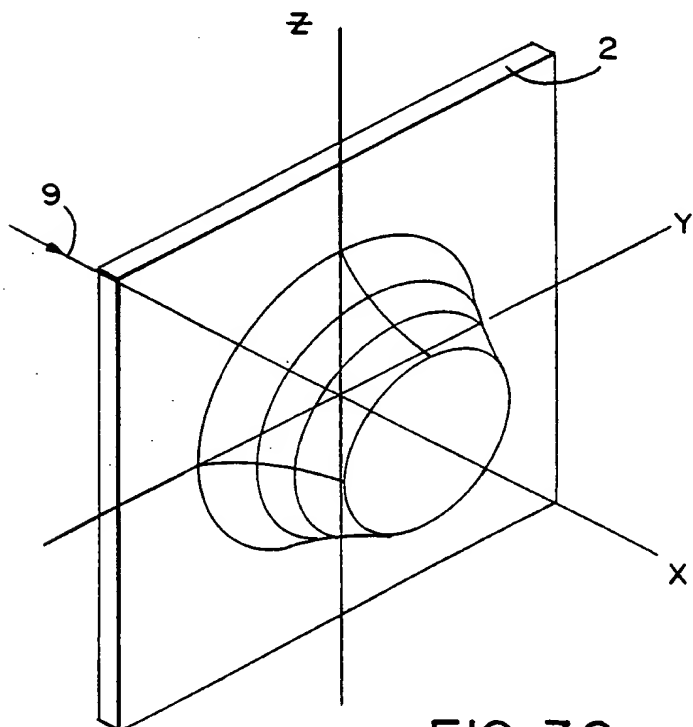
FIG.3C
PRIOR ART
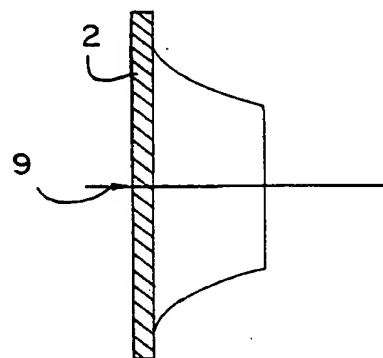
FIG.3D
PRIOR ART
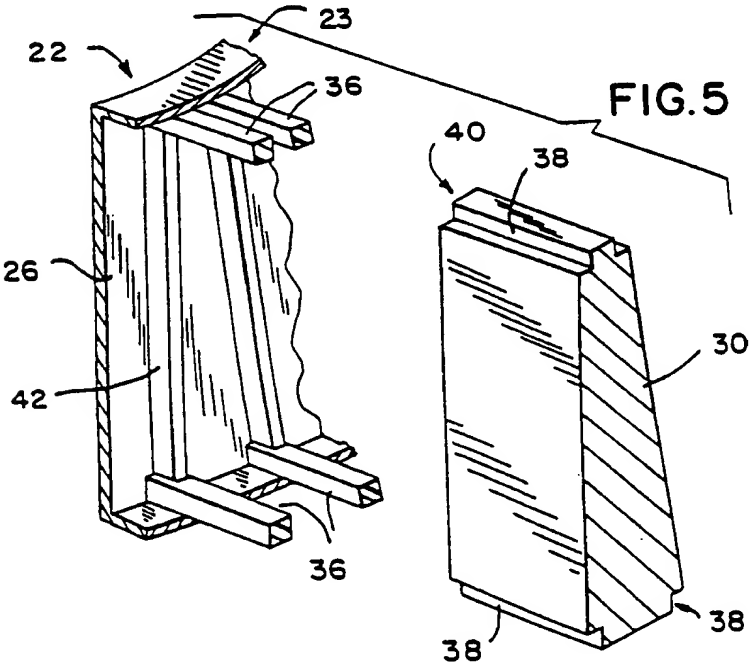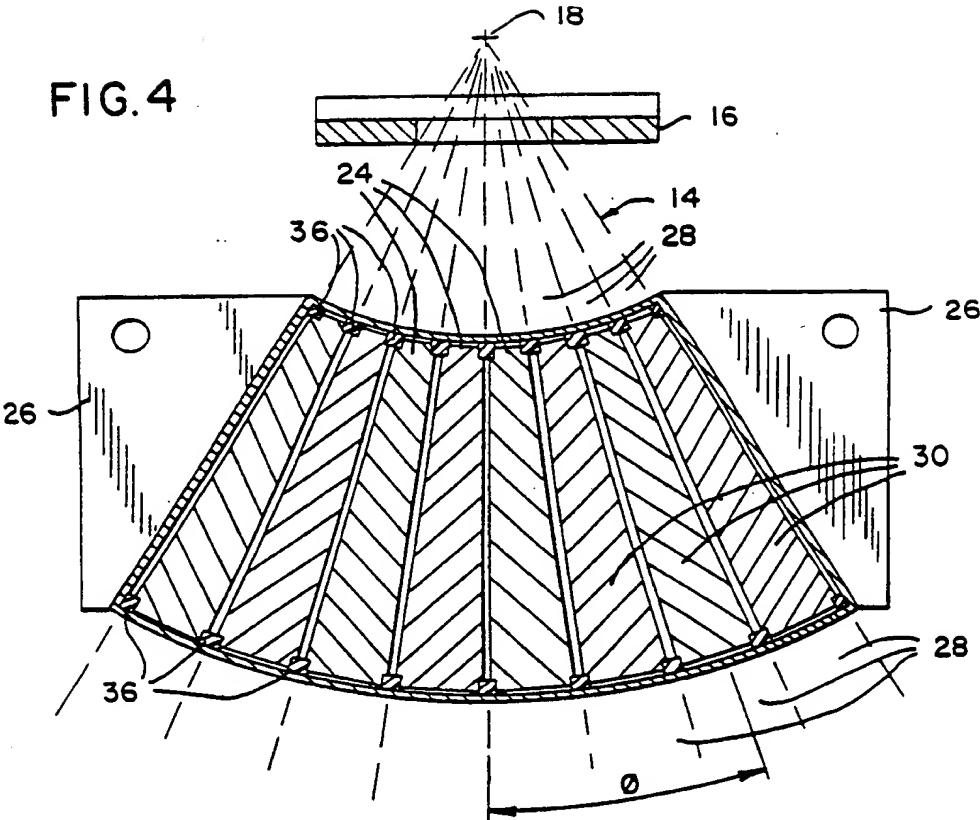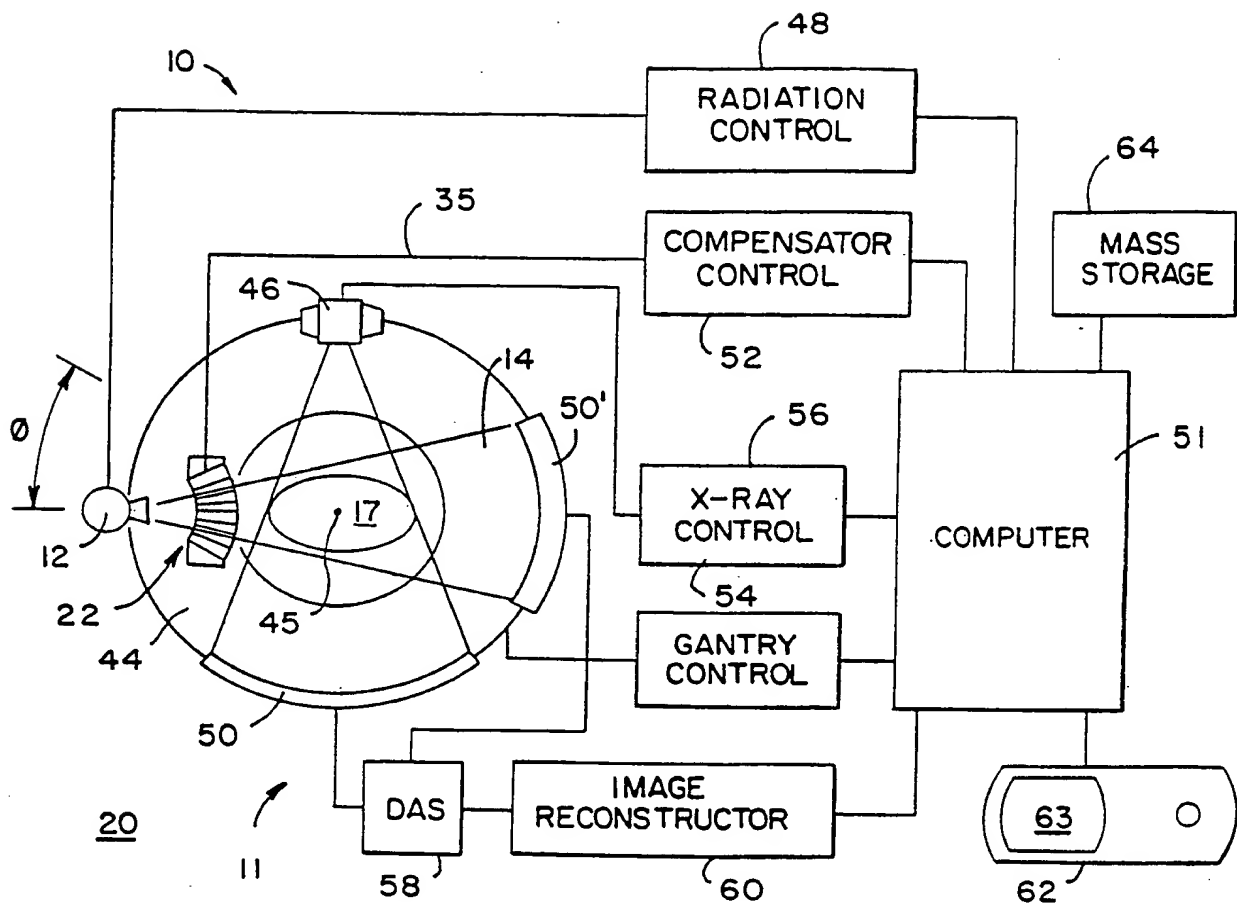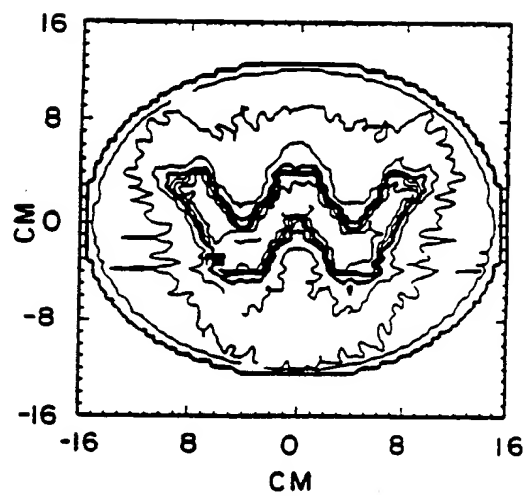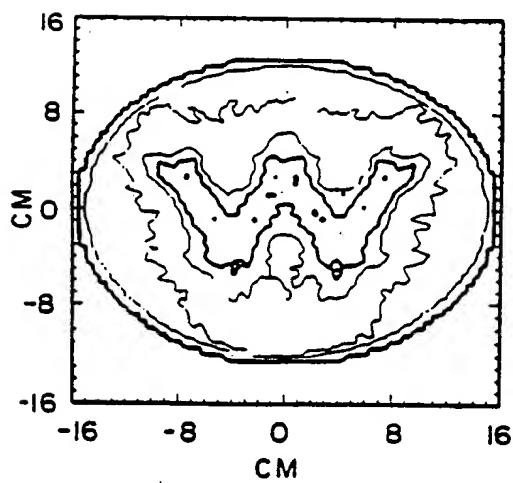
FIG.4

FIG.5

4/11

FIG. 6

FIG. 7

FIG.8a
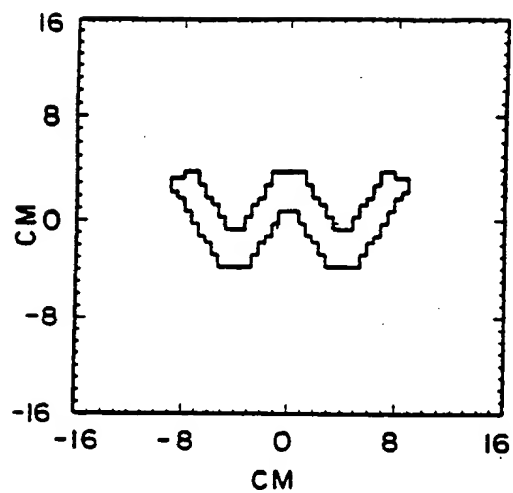
FIG.8b

FIG.8c

FIG.8d
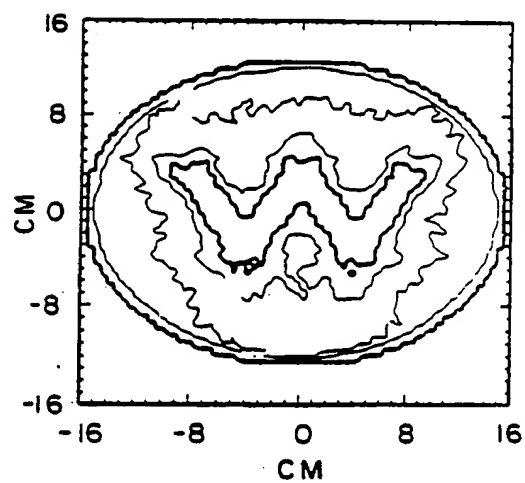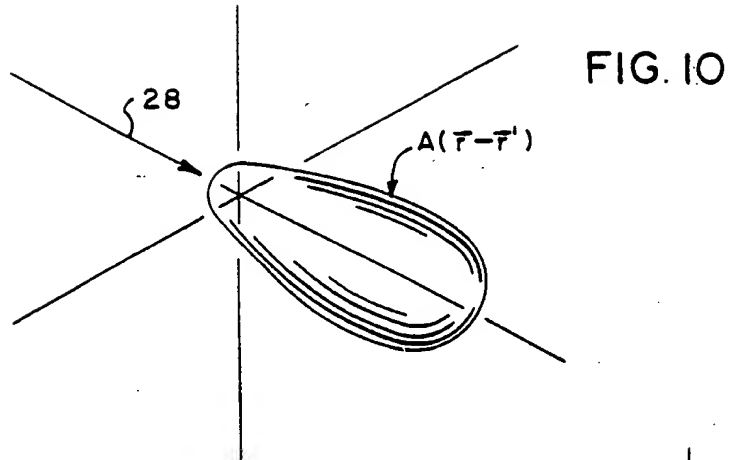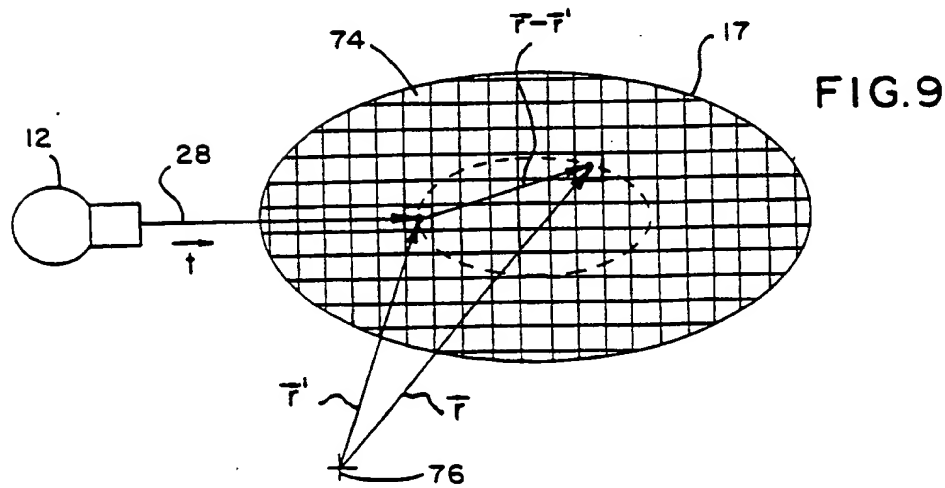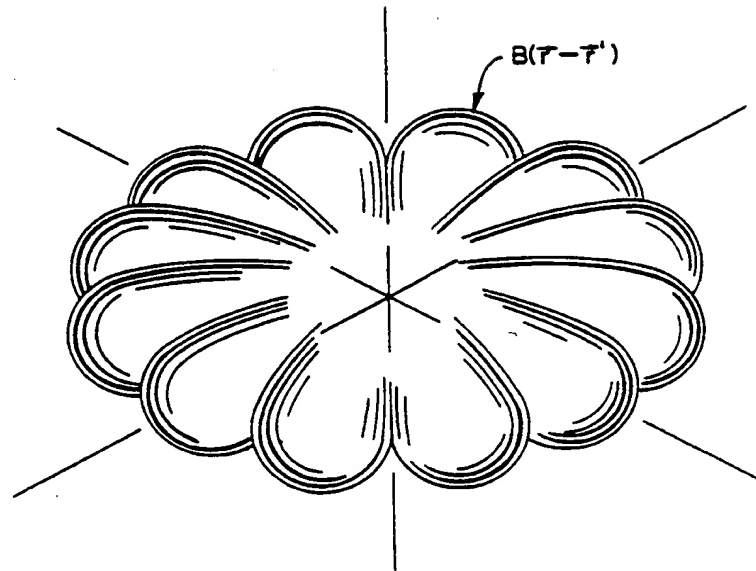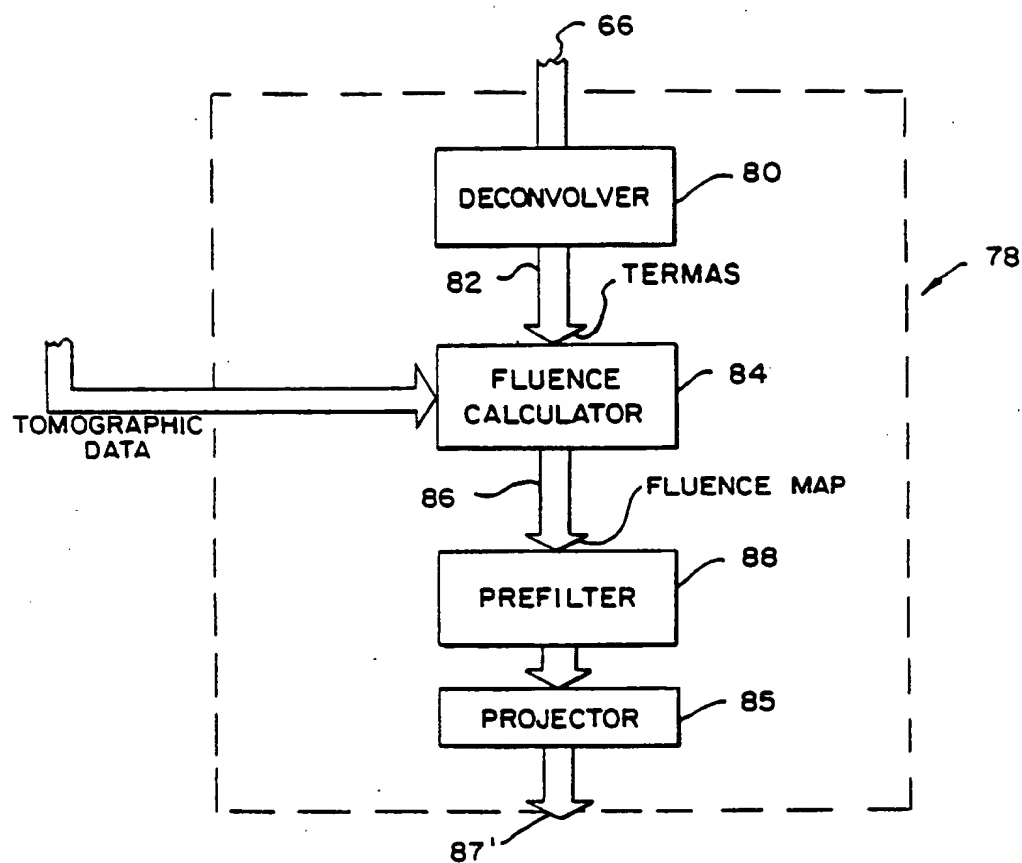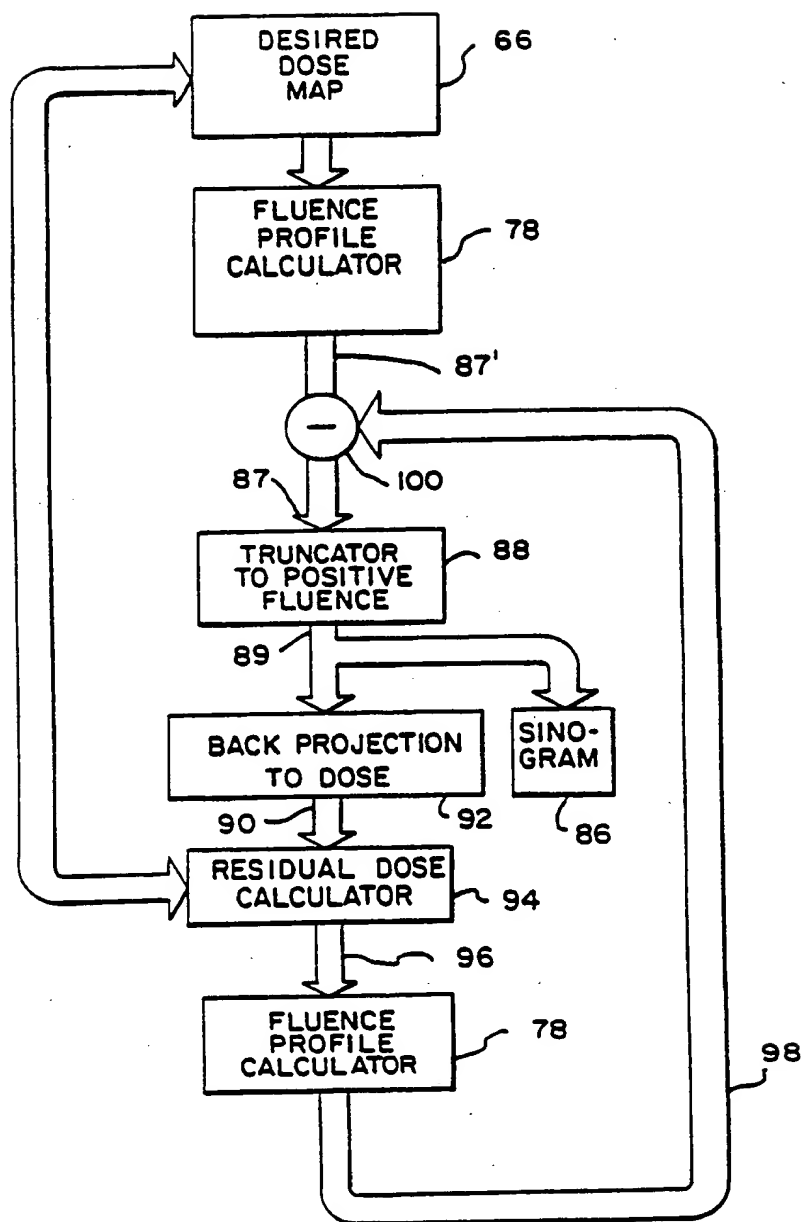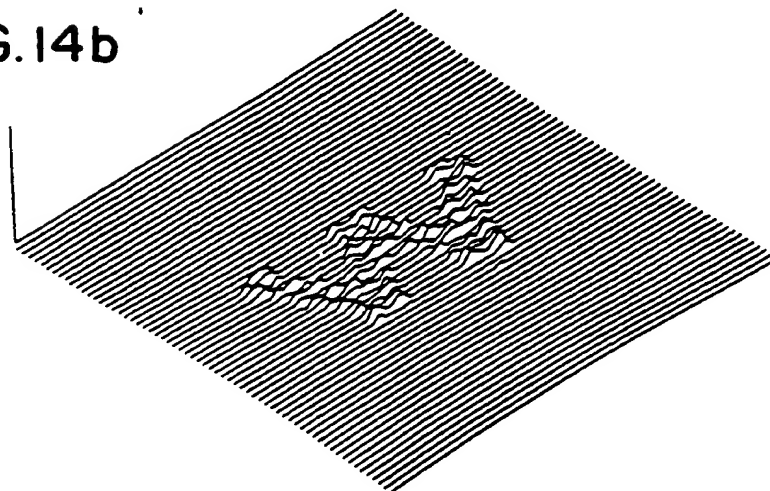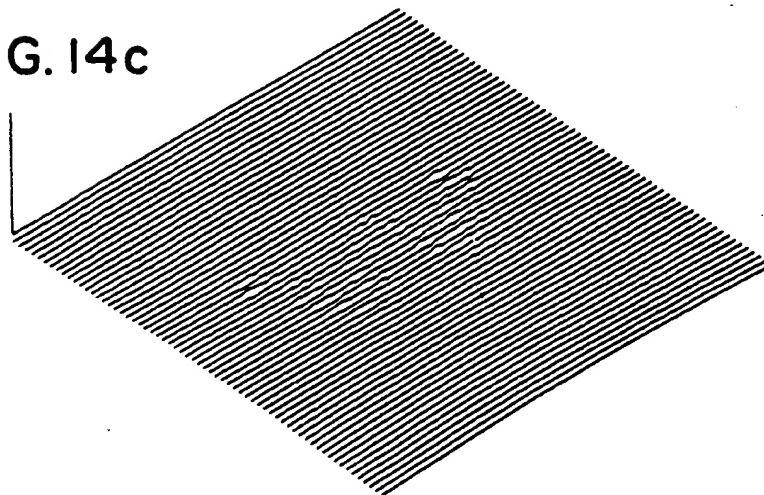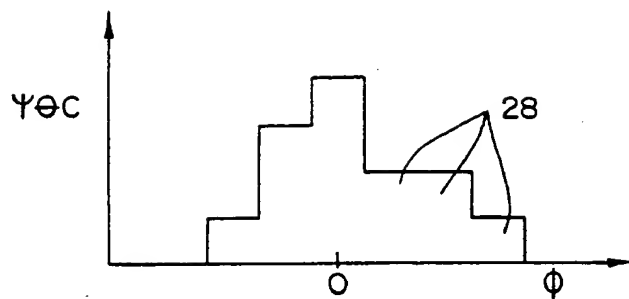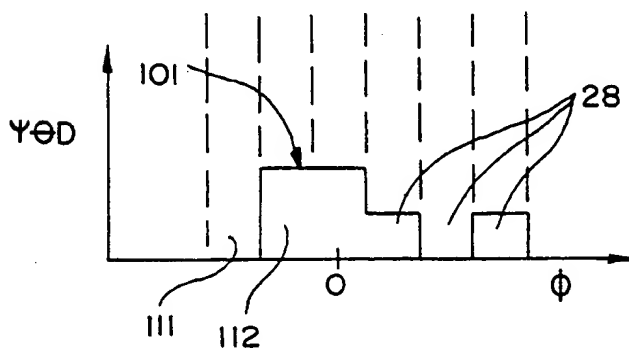
FIG. 10

28

$A(\bar{r}-\bar{r}')$

FIG. 11

$B(\bar{r}-\bar{r}')$

74 $\bar{r}-\bar{r}'$ 17

12 28 FIG. 9

$\bar{t}$

$\bar{r}'$ $\bar{r}$

76

66

```
        ┌─────────────────┐
        │   DECONVOLVER   │── 80
        └─────────────────┘
   82 ──┐        │
        │    TERMAS
        ▼        ▼
        ┌─────────────────┐
TOMOGRAPHIC │    FLUENCE    │── 84
   DATA ───▶│  CALCULATOR   │
        └─────────────────┘
   86 ──┐        │
        │    FLUENCE MAP
        ▼        ▼
        ┌─────────────────┐
        │   PREFILTER     │── 88
        └─────────────────┘
                 │
                 ▼
        ┌─────────────────┐
        │   PROJECTOR     │── 85
        └─────────────────┘
                 │
                 ▼
              87'
```

78

FIG.12

DESIRED
DOSE
MAP — 66

FLUENCE
PROFILE
CALCULATOR — 78

87'

87

100

TRUNCATOR
TO POSITIVE
FLUENCE — 88

89

BACK PROJECTION
TO DOSE

SINO-
GRAM

90

92

86

RESIDUAL DOSE
CALCULATOR — 94

96

FLUENCE
PROFILE
CALCULATOR — 78

98
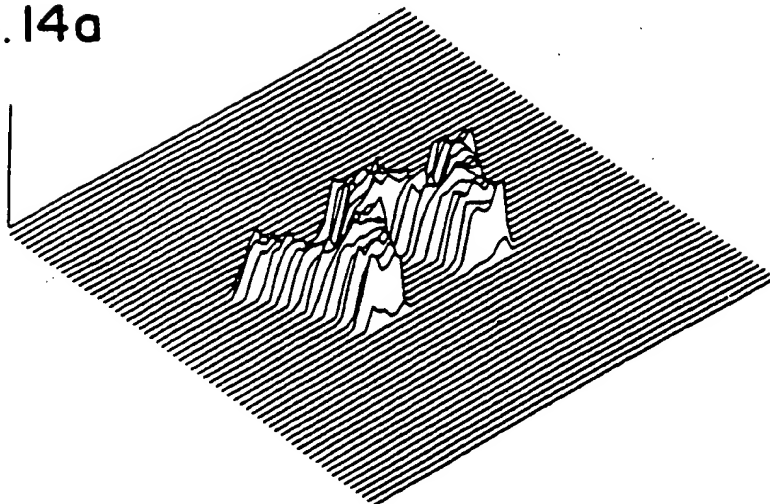
FIG. 13

FIG.14a

FIG.14b

FIG.14c

FIG. 15a

FIG. 15b

FIG. 16

FIG. 17

FIG. 18
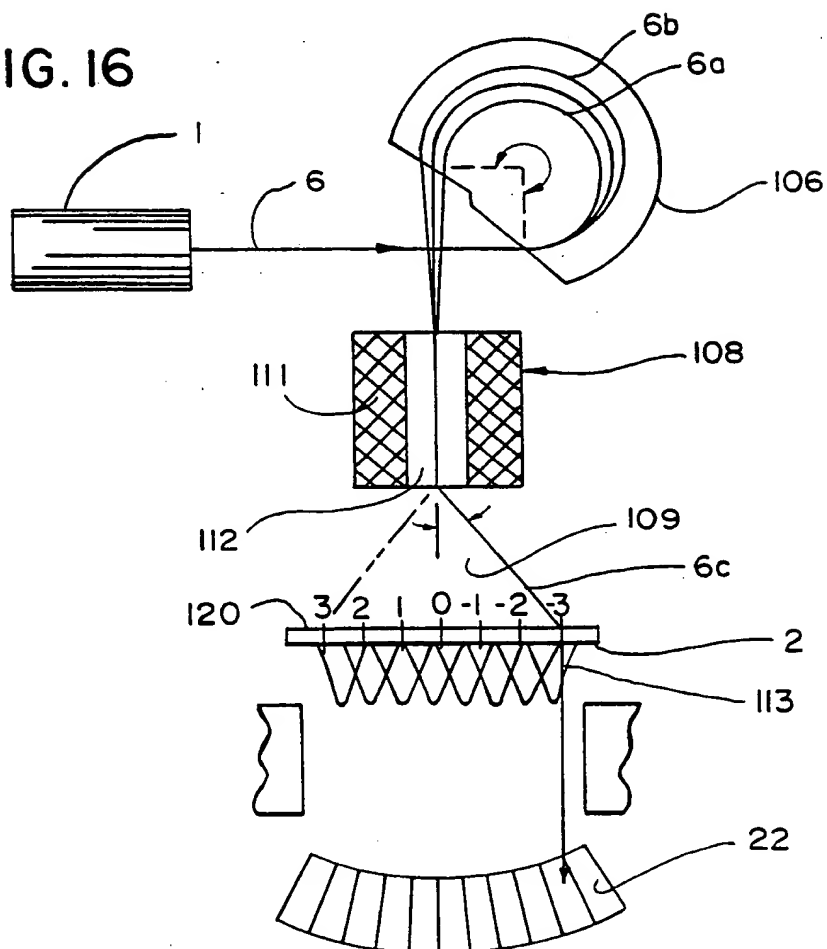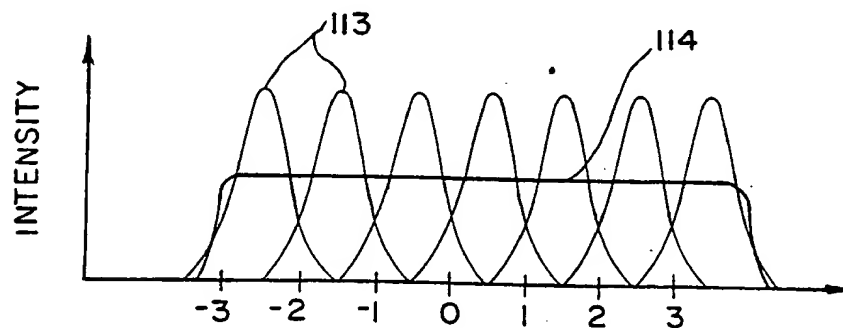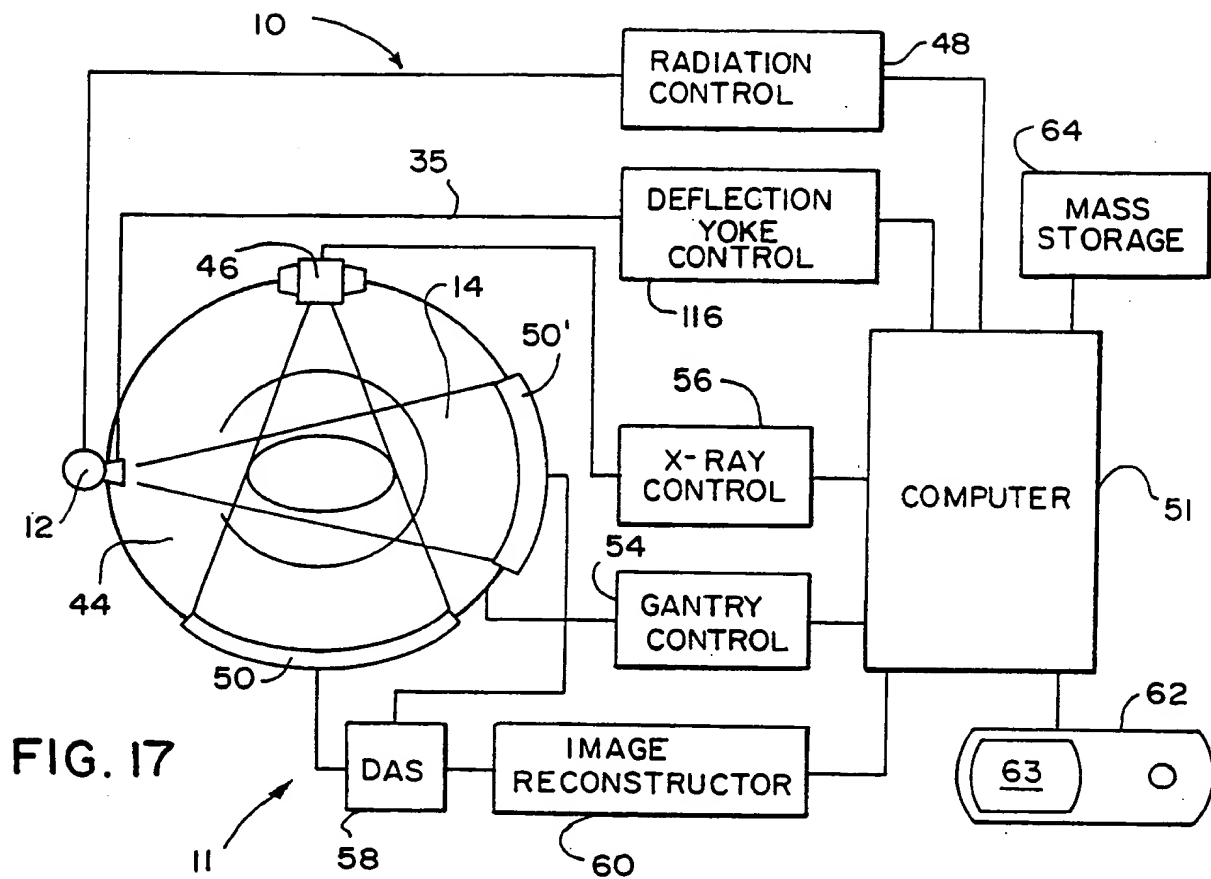
INTERNATIONAL SEARCH REPORT

A. CLASSIFICATION OF SUBJECT MATTER
IPC 5 G21K1/04 G21K1/093 G21K5/04 A61N5/10

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
IPC 5 G21K A61N

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
P,X	EP,A,0 561 533 (WISCONSIN ALUMNI RES. FOUND.) 22 September 1993	1,2
P,Y	see page 2, line 45 – page 3, line 45 see page 4, line 37 – page 5, line 9 see page 5, line 50 – page 6, line 8 see page 11, line 17 – line 38 see claims 1,2,7 ---	3
Y	FR,A,2 023 648 (PHILIPS) 21 August 1970 see page 5, line 39 – page 6, line 16 ---	3
A	EP,A,0 083 198 (ANDREX RADIATION PROD.) 6 July 1983 see page 4, line 8 – line 18 see page 4, line 29 – page 5, line 3 ---	3-6
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Date of the actual completion of the international search	Date of mailing of the international search report
26 September 1994	1 2. 10. 94

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C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

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